1972

A biomechanical investigation of the ability of three types of bone graft-vertebral body constructs to withstand axial compression

Jesse B. Jupiter
Yale University

A BIOMECHANICAL INVESTIGATION OF THE ABILITY OF THREE TYPES OF BONE GRAFT-VERTEBRAL BODY CONSTRUCTS TO WITHSTAND AXIAL COMPRESSION

JESSE B. JUPITER

1972

Permission for photocopying or microfilming of "A Biomechanical investigation of 3 types of bone graft vertebral constructs"

(TITLE OF THESIS)

Jesse Jupiter

Signature of Author

May 12, 1972

Date

A BIOMECHANICAL INVESTIGATION
OF THE ABILITY OF
THREE TYPES OF BONE GRAFT-VERTEBRAL BODY CONSTRUCTS
TO WITHSTAND AXIAL COMPRESSION

by

Jesse B. Jupiter
A. B. Brown University 1968

A thesis submitted to the
Department of Orthopaedic Surgery
Yale University School of Medicine
in partial fulfillment
of the requirements for the degree of
Doctor of Medicine

April 1972

New Haven, Connecticut

ACKNOWLEDGEMENT

To Dr. Augustus A. White III, for his guidance, patience, advice, and most of all his friendship.

To Dr. Manohar Panjabi, for his instruction, both technical and theoretical.

To Tom Auten and Michael Ricciuti, for advice and assistance in the operation of the Instron Testing Machine.

To Mrs. Mary Ann Jenkins and Miss Joan Laurie, for allowing the use of their equipment.

To Kevin Freeman and Dennis Gawitt, for their technical assistance in obtaining roentgenograms.

To Dr. Kristaps J. Keggi, for providing important material for this investigation.

And finally, to my dear wife, Beryl, for her devotion and patience in assuming the role of my typist and statistician.

DEDICATION

To My Family

TABLE OF CONTENTS

INTRODUCTION

The surgical procedure of spinal fusion is employed by surgeons for the treatment of various conditions affecting the spinal column and nervous system. Fusion involves procedures done on the bony elements of the vertebral column in order to obtain an osseous union of a specific number of its segments. Different configurations of bone grafts have been utilized in fusion procedures of the cervical and lumbar spines. This investigation is intended to study the load bearing capacities of several types of graft-vertebral body constructs.

Through an anterior surgical approach, several investigators have demonstrated the ability to perform cervical spinal fusions, using different types of autogenous iliac crest bone grafts. The first, a horseshoe-type graft, was employed by Smith and Robinson (1955) and Robinson and Southwick (1961). The second is the dowel-type graft employed by Cloward (1958), and the third is a strut-type of graft used by Hodgson and Stock (1956) and Bailey and Badgely (1960). These investigators have demonstrated that through the anterior approach fusions of the cervical spine can be achieved with greater ease, less morbidity, and with a relatively high rate of arthrodesis.

Early post-operative mechanical problems with grafts

of anterior cervical fusion including collapse of the graft, expulsion of the graft from the vertebral bodies, and collapse of the vertebrae over the graft have been noted (Crandall and Batzdorf, 1966; Jackson and Delucca, 1966; Kebish and Keggi, 1967; Aronson, 1968; Galera and Tovi, 1968; Simmons and Bhalla, 1969). Although Simmons and Bhalla (1969) have compared biomechanically the ability of a modified strut graft and a dowel graft to withstand torsional force, the major force to which the graft-vertebra construct is subjected during the early post-operative period may be one of compression. This is a result of the axial loading developed during ambulation. Although each of the principal proponents of the anterior cervical fusion technique differ in regard to how soon the patient is to be ambulated post-operatively, all describe the use of a neck brace for various lengths of time which helps reduce the torsional loading on the graft site. Thus, the early strength seems dependent upon the capacity of the graft-vertebral body construct to withstand a vertical compressive force.

White (1971) has subjected three types of grafts, the horseshoe type of Smith, Robinson, and Southwick, the dowel type proposed by Cloward, and a modified strut graft employed by Bailey and Badgely, to vertical compressive loads. The horseshoe type of graft proved to withstand

loads significantly greater than the other two types (Table 1). This investigation demonstrated, however, that all three grafts were able to withstand loads significantly higher than those occurring in the ambulatory man, with the range for the averages of the specimens approximately 2.5 to 5 times the average body weight. Thus, the weak point of the procedure probably is not the individual graft itself, but the combined graft-vertebrae construct, which includes the manner by which the vertebral bodies are changed to accommodate the graft. Thus, it was decided to investigate this problem by performing the surgical procedures on the vertebrae of fresh autopsy specimens, using autogenous iliac crest as a source for the different grafts and submitting the entire mechanical construct to axial compression.

THE HISTORY OF SPINAL FUSION

According to Bick (1964), the history of the spinal fusion as an effective surgical procedure really began with Hibbs (1911) and Albee (1911) who performed lumbar fusions through a posterior approach for tuberculosis of the spine. Their procedure was widely employed in the United States. A significant development occurred when Ghormley (1933) demonstrated the ready availability of autogenous cancellous bone from below the iliac crest for use in lumbosacral fusions. This stimulated a great deal of investigation to compare the relative effectiveness of cancellous bone to rigid cortical bone in fusion procedures.

In 1936, two reports, those of Mercer (1936) and Jenkins (1936), appeared independently in the British literature describing an anterior approach to fusion of the lumbar spine. Mercer employed bone from the iliac crest as an anterior graft between the bodies of L-5 and S-1. Jenkins fused the same level, although he employed a tibial cortical graft. These papers stimulated new interest in the anterior approach to the spine. This approach has been facilitated by advances in anesthesia, fluid replacement, and surgical technique, and is now being effectively applied to cervical spinal fusions, using the various types of grafts described in the Introduction.

PREVIOUS BIOMECHANICAL COMPRESSION STUDIES OF THE SPINE

The scientific literature includes values for more than 300 vertebrae subjected to compression studies. Siegfried Ruff (1950), in an attempt to study the capacity of the spinal column to absorb energy, subjected vertebral-disc units of up to five vertebrae to vertical central axis loading. According to his method of evaluating the load capacity, he took the "breaking load" to be that point at which the stress-strain curve had its first peak, considering this to be the point at which the vertebral body experienced the initial irreversible injury. The first complex T-10 to L-3 withstood a total load of 690 kg with a deformation of 12 mm until T-12 "broke." When the load was increased to 840 kg, T-11 and L-1 failed. Ruff also investigated the failure load for individual thoracic and lumbar vertebrae (Table 2) and determined the portion of the total body weight supported by individual vertebrae (Fig. 1b).

Olof Perey (1957) subjected individual vertebra and groups of two and three lumbar vertebrae with intervening discs to static and dynamic compression loading. His definition of the "breaking point" was the greatest value of stress which can be obtained before the material breaks. His results for each group tested showed a marked variance

of strength related to the age and individual specimen. In the experimental series on two vertebrae, for the groups "over sixty" years of age, the average breaking strength was 425 kiloponds with a range of 290 to 530 kiloponds, while the average for the group "under forty" was 780 kiloponds with a range of 510 to 1100 kiloponds. Many of the specimens showed end-plate fractures. In the experimental series of individual lumbar vertebrae, the average for the "under sixty" group was 600 kiloponds, while for the "over sixty" group the average was 260 kiloponds. He also noted that the resistance of the vertebral end-plate decreased with age. It is evident that Perey's values are lower than Ruff's, which may be accounted for by the fact that Perey loaded both the vertebral body and the articular processes, while Ruff loaded only the vertebral bodies (Fig. 2).

Brown, Hansen, and Yorra (1957) in an investigation aimed primarily at the mechanical properties of the intevertebral disc, subjected units of two lumbar vertebrae with intervening disc to axial compression. They found the ultimate static compressive load to be between 1000 and 1300 lbs. (Table 3). Besides removing the posterior elements of the vertebral bodies, these investigators sawed off the top and bottom, respectively, of the upper and lower vertebral bodies in order to obtain more parallel surfaces for axial loading (Fig. 2). They noted that the

specimens failed in a similar manner. This was characterized by audible cracks followed by the leakage of sanguinous fluid. Inspection and palpation revealed little gross abnormality. They noted extensive collapse of the bony end-plate and underlying trabeculae occurring at relatively small loads in osteoporotic specimens. They showed that the failure took place in all instances in the vertebral end-plates, even when well formed rupture of the annulus in the disc were present.

Evans and Lissner (1959) studied the intact lumbar spines of embalmed and unembalmed adult males under axial loading and bending tests. Their average maximum load for the embalmed specimens was 882 lbs with a range of 610 to 1350 lbs. For the unembalmed spines, the maximum load was 544 lbs with a range of 290 to 690 lbs (Table 4).

Roaf (1960) also conducted compression studies of several types of vertebral-disc complexes. To absorb the force of compression, he postulated a vertebral mechanism which involves initially the bulging of the end-plates, thereby causing blood to be squeezed out of the cancellous bone of the vertebral body. As the pressure is increased, the end-plate bulges more and finally cracks (Fig. 3). It is Roaf's opinion that should the line of force be directed obliquely to the end-plate, the fracture line would be oblique. Roaf also observed that the intact disc was more

resistant to vertical compression than the vertebral body.

Nachemson (1960) also tested preparations of two vertebrae with intervening disc (Fig. 2). With individuals age 20 and 22, his results were 900 to 1100 kg, while with those 46 and older, the maximum loads were 460 kg and less.

Crocker and Higgins (1966) performed static and dynamic compression tests of isolated vertebrae (Fig. 2). They noted how the vertebral bodies were exceedingly stiffer in axial compression than the intevertebral discs, by a ratio as much as 5:1.

Rockoff, Sweet, and Bleustein (1969) investigated the contribution of the cortical shell and the central trabecular bone to the peak compressive strength of the human lumbar vertebrae. They found that the cortex contributes 45-75% of the peak strength, regardless of the physical density or ash content of the trabecular bone. This finding is in agreement with Evans (1957) who believed the load supporting part of the vertebral body was the compact bone. They also noted the role of the trabecular bone in contributing to the strength of the vertebrae depends upon its ash content, so that if the ash content is less than 59%, only 40% or less of the forces are transmitted directly by the central trabecular bone. It was also noted that the bone density and ash content decrease with age. They found that over the age of 40, the peak compressive strength of vertebrae is markedly lower (Fig. 4).

From the work of these investigators, a basic understanding of the vertebral body's ability to withstand compression can be appreciated. Such factors as the amount of cortical bone, the integrity of the bony end-plate, and the age of the specimen have all been shown to contribute to the strength of the vertebral body.

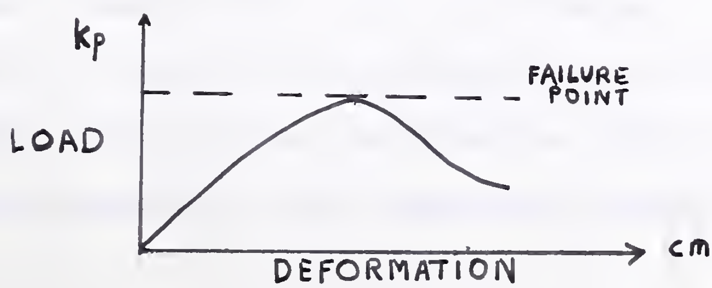
DEFINITIONS

As this investigation involves biomechanical studies, a brief description of terminology is in order. According to Perey (1957), biomechanics may be defined as a study of "those phenomena in the living body which are basically of a mechanical nature." If an object is subjected to an external force, either that of compression, shear, extension, or torsion, it will develop internal stresses and may experience changes in form, depending upon the nature of the material. By integrating the total load over the applied area, the magnitude and direction of the equivalent force vector can be determined, and this idealized force may be used in a biomechanical study of the system (Frankel and Burstein, 1970).

It is important to note that biological tissues exhibit to varying degrees what is known as viscoelastic behavior. The concept of a viscoelastic material can best be understood by construction of an idealized model. This consists of two components, an ideal Hookean body or spring and an ideal Newtonian body or syringe. The elastic spring-like part of a viscoelastic material has a time independent relationship between load and deformation; and when the applied load is removed, it returns back to its original shape. The viscous syringe exhibits a time dependent relationship with the rate of deformation a direct

function of the load. Here there is no tendency to return to the original dimension when the load is removed. As described by Frankel and Burstein (1970), if a load is allowed to remain for an appreciable period of time and is then removed, there is an immediate elastic recoil toward the material's original dimension. The material does not, however, reach its original dimension until after a definite period has elapsed. This response to prolonged loading is typical of viscoelastic material. This complex nature must be born in mind in all studies of the mechanical behavior of bone and other living tissues.

With compression, the object experiences stresses which are directed in the same direction as the applied force. Some shear stresses on planes directed obliquely to the line of application of the compressive load will undoubtedly be present. As the compressive force is applied to an object such as a vertebra, it will become shorter and wider. The breaking point for our experiment may be defined as that amount of external force at which the material can no longer withstand the increasing amount of stresses developed. This is the peak of the load-deformation curve.



MATERIALS AND METHODS

A. Materials

All the specimens used in this investigation were obtained from the pathology department of the Yale-New Haven Hospital. Cervical spines would have been ideal; however, they could not be obtained due to the resultant cosmetic alteration of the cadaver. Consequently, thoracic spines were employed. A section of either right or left anterior iliac crest was also taken from each autopsy specimen. The age, sex, and cause of death of each specimen is documented in Table 5. Specimens were not taken from patients who had any neoplastic or infectious disease which could affect the bone, or any vertebral deformity such as scoliosis, fracture, or marked osteoporosis.

B. Preparation of Materials

The entire thoracic spine, including the posterior processes was removed from the cadaver at the time of autopsy. The excised spines and respective iliac bone were cleaned of much of the soft tissue, leaving the ligaments intact, and stored at -20° until time of testing. It has been demonstrated that the freezing of bone does not affect its physical properties (Evans, 1957; Sedlin and Hirsch, 1966; Crocker and Higgins, 1966). Roentgenograms were obtained of each specimen, enabling the

thoracic level of each vertebral body to be determined and recorded.

At the time of testing, the vertebral units, consisting of two vertebrae with the intervening disc, were cut on a band saw while still in the frozen state. The disc tissue was carefully excised from the upper and lower surfaces of the vertebrae, so that these surfaces consisted of only the vertebral end-plates. These units were placed in Lactated-Ringer's solution and allowed to thaw to the ambient temperature.

C. Technical Preparation of Surgical Constructs

The bone grafts were prepared as follows. The respective section of iliac crest was also allowed to thaw to room temperature in Lactated-Ringer's solution. The Smith-Robinson horseshoe graft, to be referred to as type I, was prepared by setting the double oscillating blades of the stryker saw 7 mm apart and cutting a section of iliac crest in the range of 19-25 mm on the long axes (Figs. 5 and 6).

The dowel graft, known as type II, was obtained using the specified Cloward dowel cutter in the manner described by Cloward (1959). The graft diameter was 12 mm (Figs. 5 and 6).

The third type of graft, the strut graft described by Bailey and Badgely (1960) was taken from the iliac crest with the two blades of the stryker saw 20-22 mm apart.

Its long axis was in the range of 19-25 mm (Figs. 5 and 6).

All three grafts were then allowed to remain in Lactated-Ringer's solution at room temperature. All of this work, as well as subsequent work, was accomplished in an enclosed environment with cool steam blown over the specimens to avoid any drying of the specimens.

With preparation of the grafts completed, the next step involved the construction of the graft-vertebral body complex. Each grouping of three vertebral units were placed in a randomized sequence built around a 3X3 latin square (Winer, 1962) in order to have each type of graft-vertebrae construct done equally at the various levels of the thoracic spine (Table 6).

The type I fusion procedure involved transection of the anterior longitudinal ligament along with the removal of the intevertebral disc. The hyaline cartilage end-plate on the top and the bottom of the disc space was carefully removed, leaving the subcondral bony end-plates intact. In some instances, in order to ensure that the graft would remain parallel to the bony end-plates, the posterior aspects of these end-plates were trimmed away, as depicted in Fig. 5 (Robinson and Southwick, 1961).

Type II was prepared in accordance with the surgical technique described by Cloward (1965). Using the Cloward twist drill attached to a Hudson cranial drill-handle, a

midline drill hole was made into both vertebral bodies and intervening disc. The hole was made deep enough to allow the bone graft to be countersunk and it varied according to the depth of the graft (Fig. 5). The remainder of the disc was left intact. The graft was attached to a special graft impacter and hammered into place. Gentle manual distraction was employed to help widen the graft site for easier implantation.

For placement of the strut graft, type III, a trough was formed in the vertebral bodies by means of a small osteotome and mallet. The graft site was made in all dimensions approximately one millimeter smaller than the graft, and after trimming the graft, gentle distraction was applied to the vertebral bodies and the graft was securely wedged into place (Fig. 5).

Roentgenograms were then made of the constructs enabling the surgical technique to be evaluated.

In order to obtain parallel surfaces perpendicular to the long axis of the vertebral bodies for the axial compression, the samples were placed in a polyester based mastic, known commercially as Plastic Padding (Fig. 7). This material hardens by an exothermic chemical reaction, unaffected by the enclosed moist environment (Hirsch, 1964). The epoxy dried in approximately ten minutes, after which the top was sanded to a smooth surface. The

polyester was subjected to compression tests by means of placing a cylinder of the material whose length was three times the diameter in the Instron testing machine and it withstood forces greater than 1000 kp with minimal deformation and no visible evidence of failure.

Compression testing of the graft-vertebrae constructs was performed on the Instron testing machine at the Yale University Schools of Engineering and Geology (Fig. 8). All specimens were tested within two hours of surgery and were transported wrapped in towels soaked with Lactated-Ringer's solution. A uniform crosshead speed of 0.127 cm/min was used. The units were placed in vertical postion on a steel platform on the calibrated load-recording cell in such a manner that the vertebral body and its posterior elements were compressed (Fig. 9). The specimens were pre-loaded twice up to 100 kp of axial compression to allow for the "setting in of materials" (Rockoff and Bleustein, 1969). The actual testing was then conducted with the failure point being the point at which the specimen could no longer support an increasing load. This was the peak load of the load-deformation curve. At this point the crosshead direction was reversed and the specimen was unloaded at the same speed used for loading. As the specimen was compressed, any points at which serosanginous fluid was expressed or audible cracks occurred were

recorded, along with any visible alterations in the graft configurations. Several units were retested after storage for several days at -20° and failed to show any significant difference in results.

Post-compression, the epoxy was stripped away and roentgenograms were taken of each specimen. The units were then carefully inspected and sliced longitudinally on a band saw. The units were then cut in half across the narrowest aspect of the whole configuration and the cross-sectional area determined by plotting the area on graph paper. The peak strength of the construct was then calculated as the ratio of the peak load born by the unit to the measured cross-sectional area, in kiloponds per square cm (Rockoff and Bleustein, 1969).

RESULTS

The results of axial compression on fifteen sets of the three types of graft-vertebrae constructs are shown in Fig. 10 and Table 7. It can be seen that the type I construct, developed by Smith, Robinson, and Southwick, withstood greater axial loading than either type II or type III. The difference between the mean values of type I, 50.9 kiloponds/sq.cm., and type II, 41.6 kiloponds/sq.cm., is statistically significant at the .10 level of confidence by the student's t distribution. Furthermore, the difference between the mean values of type I and type III, 35.2 kiloponds/sq.cm., is statistically significant at the .05 level of confidence.

It was decided to record the values in terms of kiloponds. This is defined as a force (in any direction) of 9.8065 newtons and is equivalent to the weight of one-kilogram mass under standard gravity.

The visible alterations which occurred during axial compression testing are described in Table 8 and Figs. 11, 12, and 13. As the vertebrae were compressed, along with apparent decrease in the height of the vertebral bodies, sanguinous fluid was expressed from various parts of the bodies; however, this proved to bear no consistent relationship to the eventual final peak load. This can also

be said of the occasional audible "cracks" which occurred during the compression of several specimens.

With the type I construct, aside from the expected decrease in the height of the vertebral bodies due to the axial loading, the only other visible alterations occurred in three units where the graft appeared to sink into the end-plate of either the upper or the lower vertebra (Fig. 14). This is clearly evident in the longitudinal sections (Fig. 15). In the other units tested, there were no macroscopic alterations of either the graft or vertebral end-plates (Figs. 11 and 16).

The type II construct seemed to fail in most instances by the dowel graft sinking into the lower vertebra with collapse of the anterior aspects of both vertebral bodies on the graft and subsequent loss of the interspace (Figs. 12 and 17). Longitudinal sections revealed the grafts to be intact; however, collapse into the cancellous bone of the vertebrae was quite evident (Fig. 18).

The type III construct grossly seemed to fail with widening of the vertebrae and disc along the lateral margins of the graft (Figs. 13 and 19). In some instances the vertebrae were noted to collapse somewhat over and under the graft (Fig. 19) and in several other instances the graft was extruded anteriorly. Loss of cancellous vertebral bone at either end of the strut graft was

evident in some specimens (Fig. 19).

Grafts were made of the load-deformation for each specimen as depicted in Figs. 20-25.

DISCUSSION

This investigation was initiated to determine which type of graft-vertebral body construct employed in anterior cervical spinal fusion procedures could withstand the greatest vertical compressive loading. It is reasonable to assume that vertical compression is the major vector of force experienced in the early post-operative period. The exceeding difficulty in reconstructing a biomechanical phenomenon in its entirety must be noted, as such an experiment involves, along with the ever-present variables of age, body somatotype, bone strength, and congenital defects, a number of complicating factors which are not encountered in reality. From the results obtained in this investigation, however, it appears that the type I graft-vertebrae construct developed by Smith, Robinson, and Southwick, best withstands the vertical compression applied in this experiment.

Since White's results (1971) demonstrated that all three types of grafts could withstand vertical compression significantly higher than that occurring in the ambulatory man, the weakness is, therefore, in the construction of the graft-vertebrae complex. A close look at the method by which each type of graft-vertebrae construct is created may reveal further evidence why the type I construct was

stronger. As Rockoff, Sweet, and Bleustein (1969) pointed out, the cortex of the vertebral body contributed between 45% to 75% of the peak strength to axial loading, with the cancellous trabecular bone contributing to much of the rest. The cortical bone of the vertebral body includes the bony end-plate. This appears to be a critical factor in the vertebral body strength, as several investigators, including Brown et al (1957), Perey (1957), and Hardy et al (1958), showed how the peak strength reached under compression was consistent with loss of the integrity of the end-plate. With the surgical technique involved in creating the three constructs, both type III and less so type II must lose vertebral cortical bone anteriorly and the bony end-plates must be violated. With the type I construct, the only cortical bone removed is in the most posterior aspect of the upper vertebra in order to allow the horseshoe graft to lie parallel with the vertebrae.

The trabecular system of the cancellous bone also plays a role in the vertebral body's strength, as, according to Roaf (1960), the fluid in the spongy bone may exert a hydraulic force when pressure is applied (Fig. 3). Here again, the weakest construct, type III, lost the greatest mass of the spongy trabecular bone. Type II also lost a considerable amount, while the spongy trabecular bone of type I was left intact. As the constructs were subjected

to compression, the anterior collapse of the vertebral bodies over the grafts of type II and type III was quite apparent. This may be due, in part, to the fact that the grafts were resting on the weaker cancellous bone, while the type I graft rested on the stronger bony end-plate.

Kebish and Keggi (1967) presented an excellent description of some of the mechanical problems seen clinically with the type II dowel graft- vertebrae construct. Out of 40 anterior cervical fusion procedures studied, clinically unsatisfactory results occurred in 52.5% with mechanical problems occurring in 47.5%. These problems included anterior collapse, malposition of the graft, and graft instability. His description of several instances of anterior collapse of the vertebral bodies on the hard dowel with concommittant settling of the graft in the cancellous bone and resultant narrowing of the disc space (Fig. 26) shows a striking resemblance to the results obtained in this investigation (Figs. 18 and 27).

Another problem, evident in Kebish and Keggi's work (Fig. 28), is the fact that due to the anatomic configuration of the iliac bone, the lateral cortical margins of the dowel graft may not be parallel. This occurred in several specimens tested in this investigation (Fig. 29). This presents a new biomechanical construct, as the cortical edges are not parallel, leaving some cancellous bone not

supported by the graft. This also occurred in several instances with the type III graft. While the type I graft only involves a piece of iliac crest about 7 millimeters in length, the longer rectangular strut graft includes iliac crest twenty to twenty-five millimeters in length, thus making it more difficult to find a section of the donor site where the lateral margins of the iliac crest are reasonably parallel.

According to Simmons and Bhalla (1969), in theory, to avoid extrusion, a graft should be placed as close as possible to the posterior part of the vertebral body. This was more easily accomplished with the type I graft than either the type II or type III and may help explain the absence of extrusion with the type I construct.

A comparison between the peak loads under compression obtained in this investigation and those cited in the introduction demonstrates failure at considerably lower values. This may be the result of several factors. One major variant is the age of the specimens. Previous investigators clearly showed that one would obtain considerably lower values with specimens over forty years of age. As the average age in this study was sixty-six years, this may account for the lower values. A second factor is the surgical procedures involved which led to weakening of the vertebral units. The weakest unit proved to be that which had the most vertebral architecture interfered with.

A third variant was the fact that the posterior articulating processes were also placed under compression in this investigation. Other stresses such as shear may have resulted from compression of these structures as they lie in a downward and backward inclination. It should be kept in mind that these conditions, along with the effects of the surrounding musculature, would be present in vivo. In conjunction with this, the factor of an intrinsic compressive force, as described by Nachemson (1966), should be mentioned. He devised a method of determining the intrinsic compressive force acting on the spine under various stable equilibrium conditions and found this force to be approximately three times the body weight at the same levels (Table 1A). Thus, in reality, there may be a static preload on the vertebrae-graft configuration. While this may be minimal in the cervical region one should still note its presence.

Lastly, mention should be made of the fact that thoracic vertebrae were employed, while the graft dimensions used were those for cervical vertebrae. This may have created a situation, especially with the type II and type III units, where because of the more anterior position of the graft in the thoracic vertebrae, anterior collapse may have occurred to a greater degree than if cervical vertebrae were employed. While this may be so, comparisons with the

clinical results of Kebish and Keggi seem to indicate that such situations as anterior collapse occur in cervical vertebrae with these grafts.

Thus, as Perey (1957) points out, the elasticity curves of his axial loading experiments demonstrate that the vertebral body is not a unit but a construct, which implies that if part of this system is altered, from a biomechanical point, a new body with new mechanical properties has appeared. This investigation has shown that in three types of methods developed to creat an arthrodesis between two vertebral bodies with an iliac bone graft, the horseshoe type graft developed by Smith, Robinson, and Southwick proved biomechanically to be the strongest graft-vertebrae construct.

SUMMARY

Three types of iliac bone graft-vertebral body constructs employed in anterior cervical spinal fusion were tested for their ability to withstand vertical compressive loads. The construct employing the horseshoe type of bone graft proved to withstand loads significantly higher than the constructs involving the dowel-type and modified strut-type bone grafts. These findings are significant because in the early post-operative period when the patient begins to ambulate, a major force on the graft site is vertical compression.

BIBLIOGRAPHY

Albee, F. H. Transplantation of a Portion of Tibia into the Spine for Pott's Disease. J. American Medical Association, 57:885, 1911.

Aronson, N., Filtzer, D.L., and Bagan, M. Anterior Cervical Fusion by the Smith-Robinson Approach. J. Neurosurgery, 29:397, 1968.

Bailey, R. W. and Badgely, L. E. Stabilization of the Cervical Spine by Anterior Fusion. J. Bone and Joint Surgery, 42A:565-594, 1960.

Bick, E. An Essay on the History of Spine Fusion Operations. Clinical Orthopaedics and Related Research, 35:9, 1964.

Brown, T., Hansen, R. J., and Yorra, A. J. Some Mechanical Tests on the Lumbosacral Spine with Particular Reference to the Intervertebral Disc. J. Bone and Joint Surgery, 39A:1135, 1957.

Cloward, R. B. The Anterior Approach for Removal of Ruptured Cervical Discs. J. Neurosurgery, 15:602-617, 1958.

Cloward, R. B. The Anterior Surgical Approach for Removal of Ruptured Cervical Intervertebral Discs and Vertebral Body Fusion. Instrumentation and Procedure Bulletin, Codman and Shurtleff, Inc., #1001:649-652, 1965.

Crandall, P. H. and Batzdorf, V. Cervical Spondylotic Myelopathy. J. Neurosurgery, 25:57-66, 1966.

Crocker, J. and Higgins, L. S. Phase IV - Investigation of Strength of Isolated Vertebrae. Nasa #NASw-1313, Technology Incorporated, 1966.

Evans, F. G. Stress and Strain in Bones, Their Relation to Fractures and Osteogenesis, Springfield, Illinois: Thomas, 1957.

Evans, F. G. and Lissner, H. R. Biomechanical Studies on the Lumbar Spine and Pelvis. J. Bone and Joint Surgery, 41-A:278-290, 1959.

Frankel, V. H. and Burstein, A. H. Orthopaedic Biomechanics, Philadelphia: Lea and Ferbiger, 1970.

Galera, R. and Tovi, D. Anterior Disc Excision with Interbody Fusion in Cervical Spondylotic Myelopathy and Rhizopathy. J. Neurosurgery, 28:305-310, 1968.

Ghormley, R. K. Low Back Pain, with Special Reference to the Articular Facets with Presentation of an Operative Procedure. J. American Medical Association, 101:1773, 1933.

Hardy, W. G., Lissner, H. R., Webster, J. E., and Gurdjian, E. S. Repeated Loading Tests of the Lumbar Spine. Surgical Forum, 9:690, 1958.

Hibbs, R. An Operation for Progressive Spinal Deformities. New York Medical Journal, 93:1013, 1911.

Higgins, L., Schnall, R., Brown, H., Howard, R., and Brockway, J. The Investigation of Vertebral Injury Sustained During Aircrew Ejection, Phase 2a. Nasa #NASw-1313, Technology Incorporated, 1967.

Hirsch, C. Method of Stabilizing Autopsy Specimens in Biomechanical Experiments. Acta Orthop. Scand., 34:374-378, 1964.

Hodgson, A. R. and Stock, F. E. Anterior Spinal Fusion. British J. of Surgery, 44:226, 1956.

Jackson, F. E. and DeLucca, H. The Treatment of Chronic Complications of Cervical Spine Injuries. Military Medicine, 131:403-420, 1966.

Jenkins, J. A. Spondylolisthesis. British J. of Surgery, 80, 1936.

Kebish, P. A. and Keggi, K. J. Mechanical Problems of the Dowel Graft in Anterior Cervical Fusion. J. Bone and Joint Surgery, 49-A:198-199, 1967.

Mercer, W. Spondylolisthesis, With a Description of a New Method of Operative Treatment and Notes of Ten Cases. Edinburgh Medical Journal, 43:545, 1936.

Nachemson, A. The Load on Lumbar Discs in Different Positions of the Body. Clinical Orthopaedics, 45:107-122, 1966.

Perey, O. Fracture of the Vertebral End-Plate in the Lumbar Spine, An Experimental Biomechanical Investigation, Acta Orthop. Scand. Suppl., 25, 1967.

Roaf, R. A Study of the Mechanics of Spinal Injuries. J. Bone and Joint Surgery, 42-B:810, 1960.

Robinson, R. A. and Smith, G. W. Anterolateral Cervical Disc Syndrome. Bull. Johns Hopkins Hospital, 96:223-224, 1955.

Robinson, R. A. and Southwick, W. O. Recent Advances in Surgery of the Cervical Spine. Surgical Clinics of North America, 41:1661-1683, 1961.

Rockoff, S. D., Sweet, E., and Bleustein, J. The Relative Contribution of Trabecular and Cortical Bone to the Strength of Human Lumbar Vertebrae. Calc. Tissue Research, 3:163-175, 1969.

Ruff, S. Brief Acceleration, Less Than One Second. German Aviation Medicine, World War II, Washington 25, D.C.: U. S. Government Printing Office, 1:584, 1950.

Sedlin, E. D. and Hirsch, C. Factors Affecting the Determination of the Physical Properties of Femoral Cortical Bone. Acta Orthop. Scand., 37:29-48, 1966.

Simmons E. H. and Bhalla, S. K. Anterior Cervical Discectomy and Fusion (with note on Discography: Technique and Interpretation of Results by W. P. Buff). J. Bone and Joint Surgery, 51-B:225-237, 1969.

Smith, G. W. and Robinson, R. A. The Treatment of Certain Cervical Spine Disorders by Anterior Removal of the Intervertebral Disc and Interbody Fusion. J. Bone and Joint Surgery, 40-A:607-624, 1958.

White, A. A. and Hirsch, C. An Experimental Study of the Immediate Load Bearing Capacity of Some Commonly Used Iliac Bone Grafts. Acta Orthop. Scand. 42, 1971.

Winer, P. Statistical Principles in Experimental Design, New York: McGraw Hill Co., 1962.

FAILURE POINTS IN KILOPONDS

SUBJECT \ TYPE	I-A	I-B	II-A	II-B	III
1	480	550	250	140	300
2	360	300	199*	75	260
3	590	475	380	180	255
4	175	140	100	70	90
5	355	330	140	90	235
6	195	200	190	125	140
7	390	390	250	40	165
8	360	380	130	90	220
9	380	340	220	220	120
10	280	220	125	60	165
mean	365.5	332.5	198.4	109.0	195.0

*This was a missing value substituted for by statistical analysis.

TABLE 1. White's Results of Compression of Three Types of Iliac Bone Graft

BREAKING STRENGTH OF VERTEBRAE, Kgs

	Age									
Vertebra	19	21	21	23	33	36	38	43	44	46
Th8		640		540		600				
Th9				610		720		700		
Th10		800			660	770			730	
Th11	750		720					860		755
Th12		900	690		800		800			
L1	720		840					900	800	800
L2		990			800		830			
L3	900							940		1100
L4					1100		900		950	
L5		1020						1000		1200

TABLE 2. Ruff's Results

BODY	LOCATION	LOAD (lbs)
A	L2-3	1100
	L4-5	1000
B	L3-4	1200
	L5-S1	1250
C	L4-5	1300

TABLE 3. Results of Brown, Hansen, and Yorra (1957)

SPECIMEN	AGE	SPECIMEN CONDITION	PARTS TESTED	MAXIMUM LOAD (lbs)	DEFLECTION (inches)
1	85	embalmed	L1-L5	738	0.75
2	60		L2-L5	1350	1.30
3	57		L1-L5	1002	0.90
4	65		L1-L5	890	0.86
5	82		L1-L5	978	1.21
6	73		L1-L5	625	1.45
7	47		L1-L5	610	1.77
8	60		L1-L5	862	1.31
9	58	unembalmed	T12-L5	640	2.38
10	37	unembalmed	T12-L5	652	1.23
11	51	unembalmed	T12-L5	290	1.06

TABLE 4. Results of Evans and Lissner (1959)

SUBJECT	AGE	SEX	THORACIC LEVEL	CAUSE OF DEATH
1	73	M	T3-T8	Myocardial Infarction
2	60	M	T3-T8	Perforated Appendix
3	90	M	T4-T10	Rectal Bleeding
4	54	M	T4-T10	CVA
5	53	M	T1-T6	Myocardial Infarction
6	65	M	T3-T8	Cardiac Failure
7	67	M	T2-T7	Myocardial Infarction
8	51	F	T2-T7	Ruptured Aortic Aneurysm
9	61	F	T2-T7	Myocardial Infarction
10	66	F	T4-T9	CVA
11	76	F	T2-T7	Cardiac Failure
12	69	F	T1-T7	Cardiac Failure
13	83	F	T2-T7	Pulmonary Embolism
14	58	M	T1-T6	Myocardial Infarction
15	63	M	T1-T7	Bronchopheumonia

TABLE 5. Autopsy Specimens

	HIGH	MIDDLE	LOW
GROUP 1	III	I	II
GROUP 2	I	II	III
GROUP 3	II	III	I

TABLE 6. 3 X 3 Latin Square

FAILURE POINTS IN KILOPONDS PER SQUARE CM

Subject	Type I	Type II	Type III
1	70	52	33
2	110	88	82
3	27	19	17
4	50	34	26
5	67	40	39
6	51	53	56
7	46	30	21
8	40	38	31
9	59	40	32
10	26	28	26
11	22	23	28
12	38	65	33
13	38	27	19
14	68	53	40
15	53	38	46
Mean	50.9	41.6	35.2

TABLE 7. Results of Compression of the Three Types of Graft-Vertebral Body Constructs

TABLE 8
VISIBLE EVIDENCE OF COMPRESSION*

UNIT NO.	TYPE I	TYPE II	TYPE III
1	None	Configuration widened around graft; graft forced outward	Lateral margins of construct around graft widened
2	None	Anterior collapse of vertebra over graft	None
3	Graft pushed into lower vertebra	Graft extruded anteriorly with collapse of graft-vertebral space	Graft extruded anteriorly
4	Cortical margins of graft failed	Anterior collapse of construct over graft; graft appeared to sink into lower vertebra	Graft appeared to sink into lower vertebra with widening of lateral margins of construct
5	None	None	Widening of lateral margins
6	Graft seemed to be pushed into lower vertebra	Anterior collapse of construct over graft	Widening of lateral margins; graft pushed out anteriorly
7	None	Graft extruded anteriorly	None
8	None	None	Configuration widened; graft appeared to sink into lower vertebra
9	None	Anterior collapse of construct over graft; widening of graftsite	Lateral margins widened around graft

UNIT NO.	TYPE I	TYPE II	TYPE III
10	Graft appeared to sink into lower vertebra	Widening of graft site; graft seemed pushed anteriorly	None
11	None	Widening of graft site; anterior collapse over graft	Graft pushed out partially
12	None	None	None
13	None	Anterior collapse over graft	Collapse of construct over graft
14	None	Widening of graft site	None
15	None	None	Lateral margins widened around graft

*In all instances as a result of axial loading, the vertebral bodies were compressed in height and increased in horizontal width with expression of sanguinous fluid.

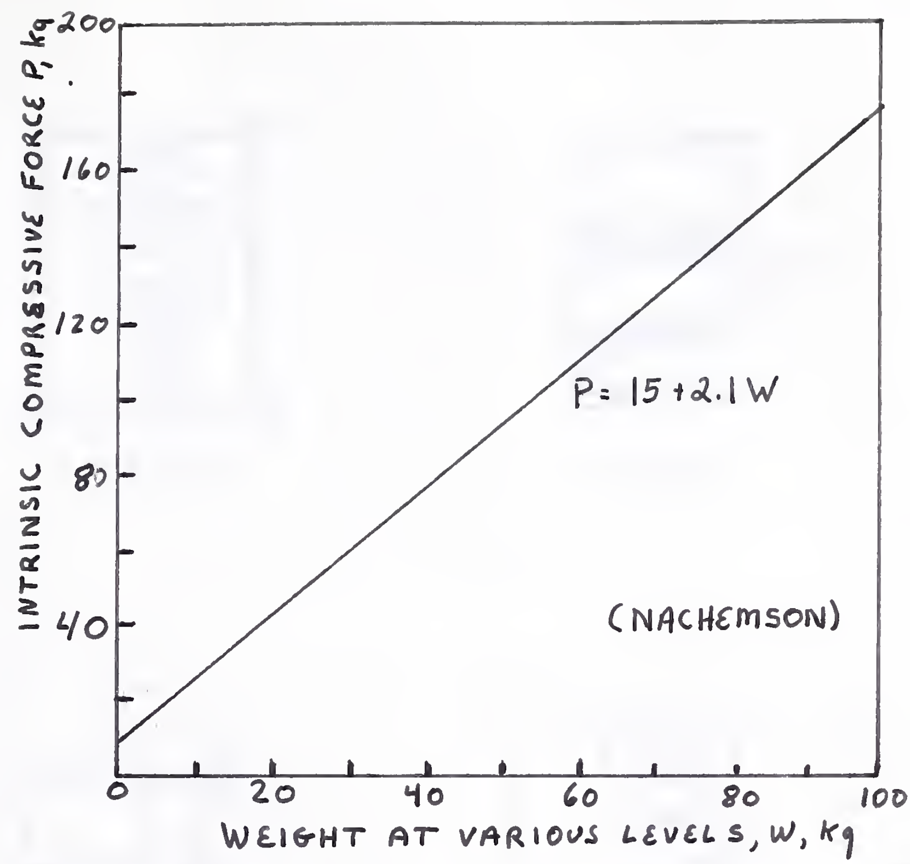


Fig. 1A. Graph of Intrinsic Compressive Force at Various Levels of the Spine (Nachemson, 1966)

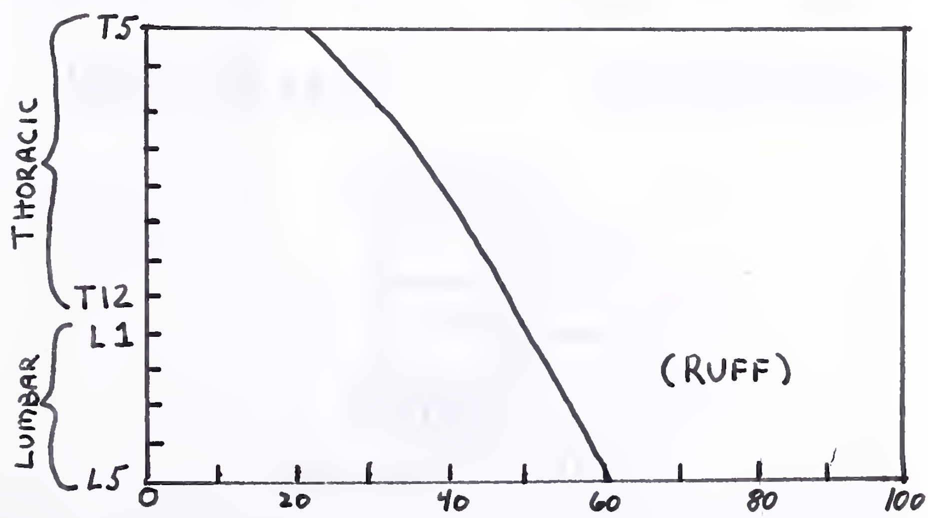


Fig. 1B. BODY WEIGHT ABOVE VARIOUS LEVELS IN VERTEBRAL COLUMN

Fig. 2. Methods of Loading Specimens
(adapted from Higgins et al, 1967)

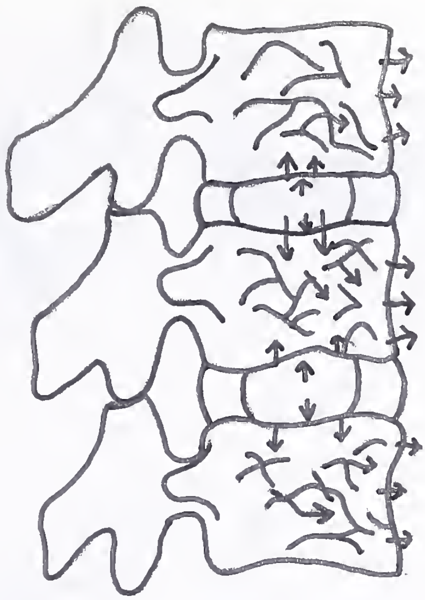


Fig. 3. Roaf's Model of the Spinal Mechanism to Absorb Compressive Force (adapted from Roaf, 1960)

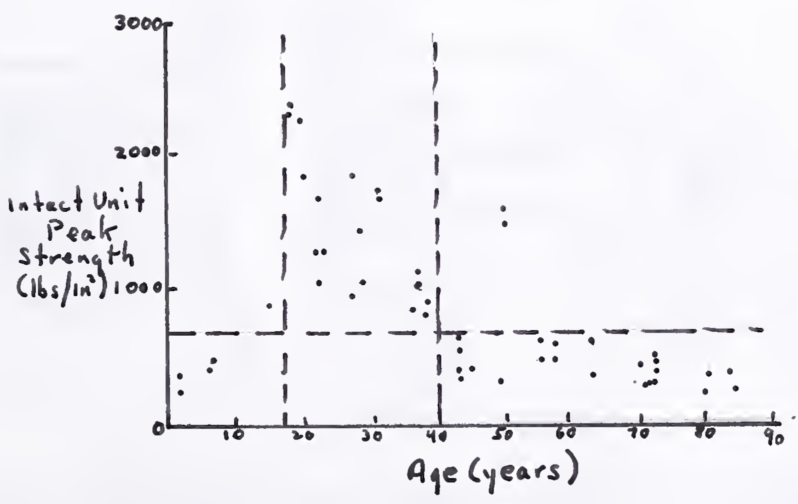


Fig. 4. Peak strength of intact units as a function of age of cadavers from which they came (adapted from Rockoff, Sweet, & Bleustein, 1969)

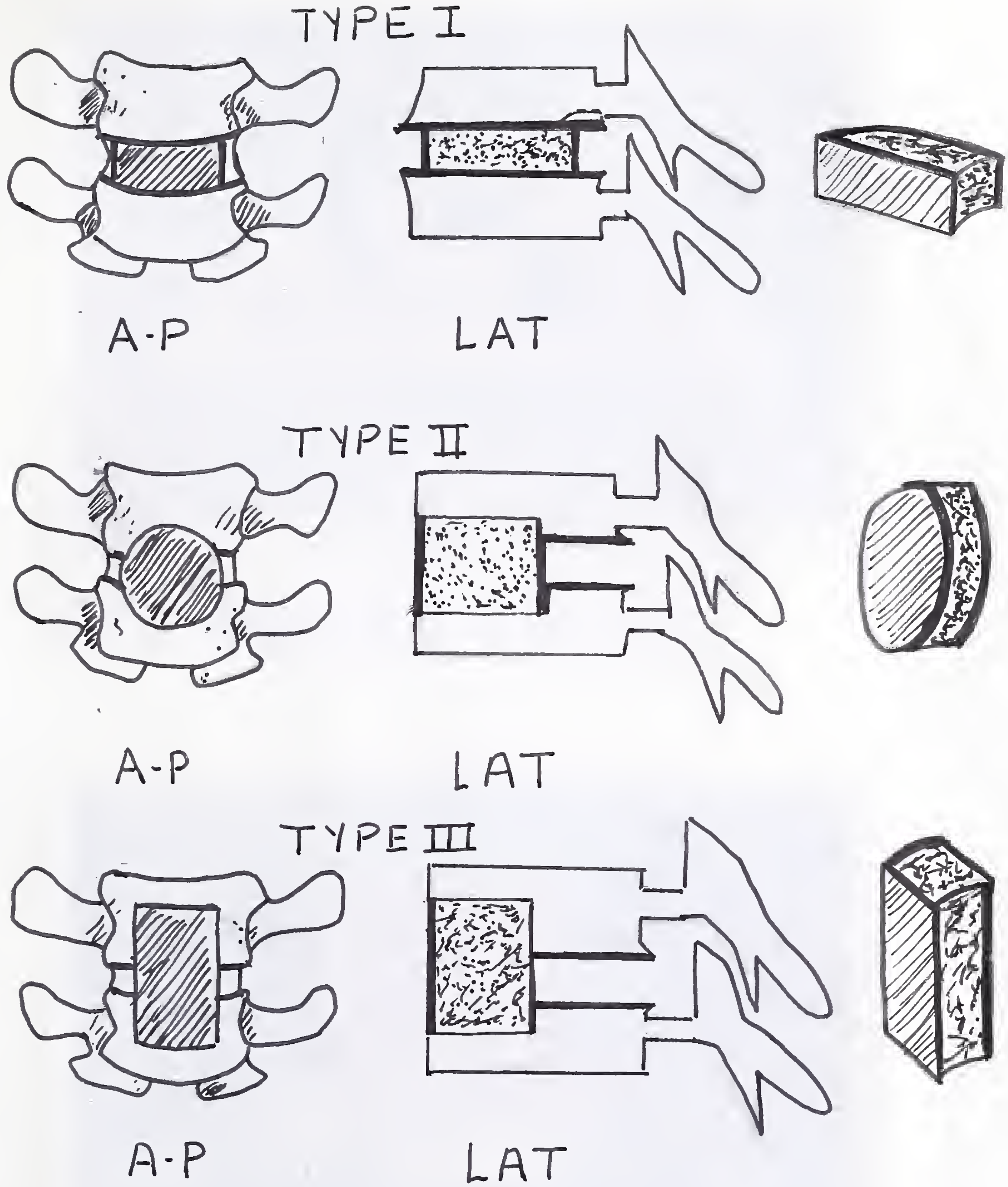


Fig. 5. Diagrams of the Three Types of Graft-Vertebral Body Constructs

Fig. 6. Section of Iliac Crest

Fig. 7. Specimen in Plastic Padding

Fig. 8. Instron Testing Machine

Fig. 9. Specimen Mounted on Calibrated Load Cell

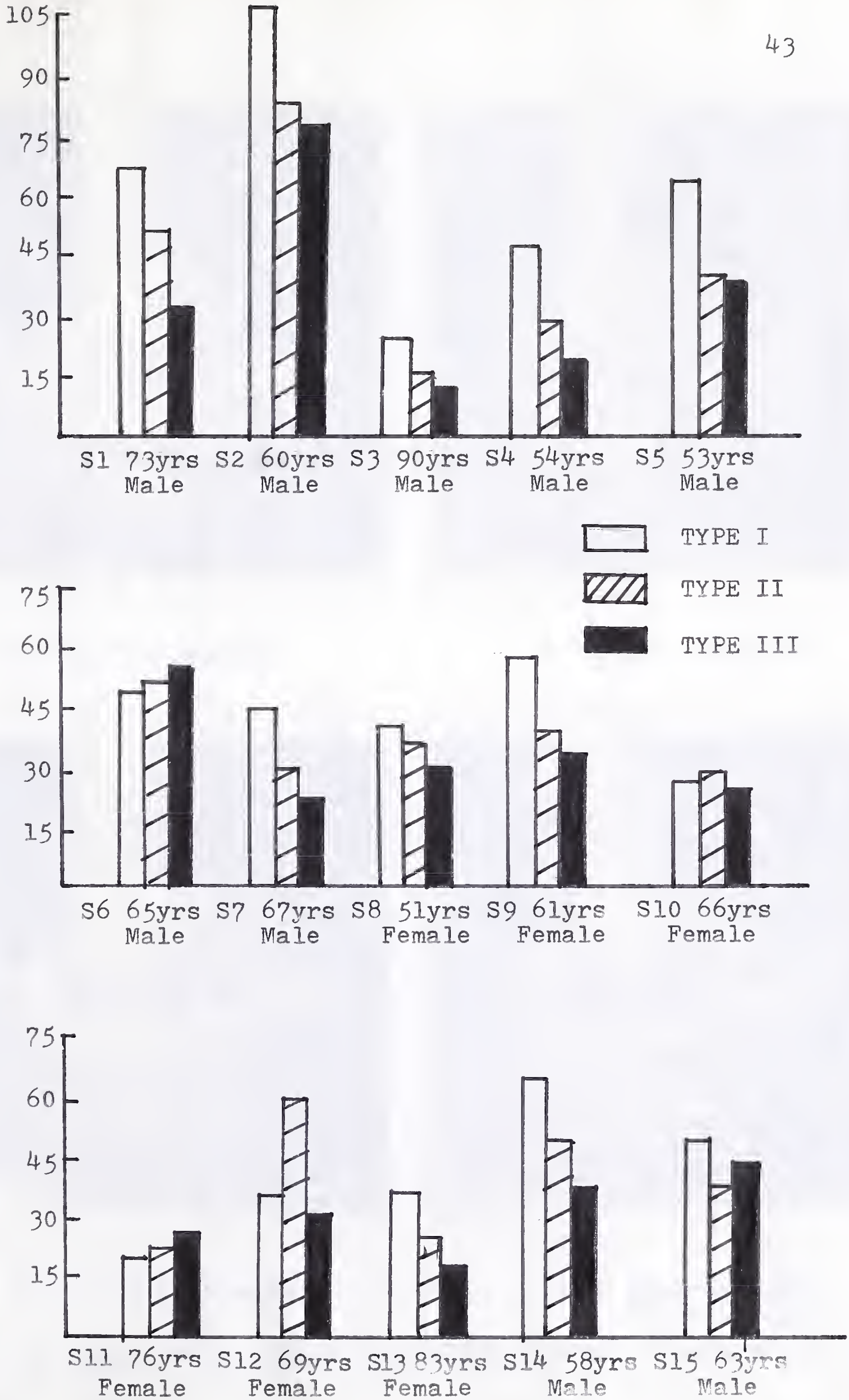


Fig. 10. Histogram of Failure Point in Kiloponds/Sq.Cm. for the Different Type Graft-Vertebrae Constructs

A. At 50 kiloponds

B. At 150 kiloponds

C. At 250 kiloponds

D. At failure point,
440 kiloponds

Fig. 11. Type I Graft-Vertebrae Construct Under Compression

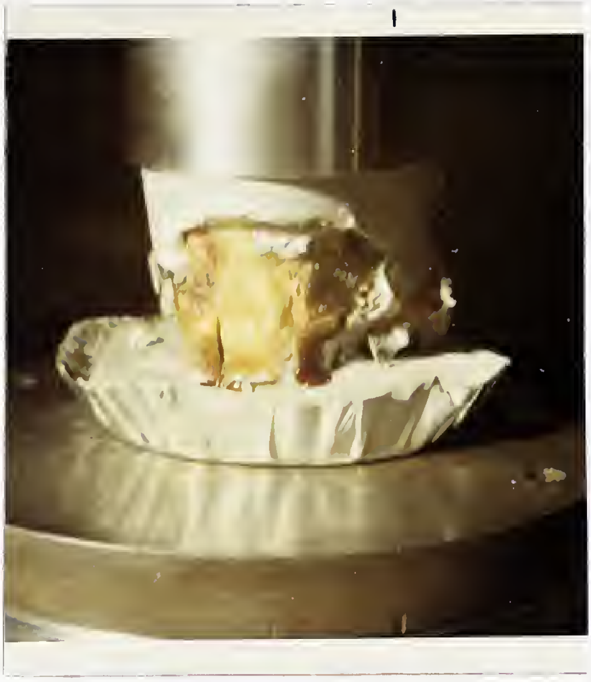
A. At 100 kiloponds

B. At 200 kiloponds

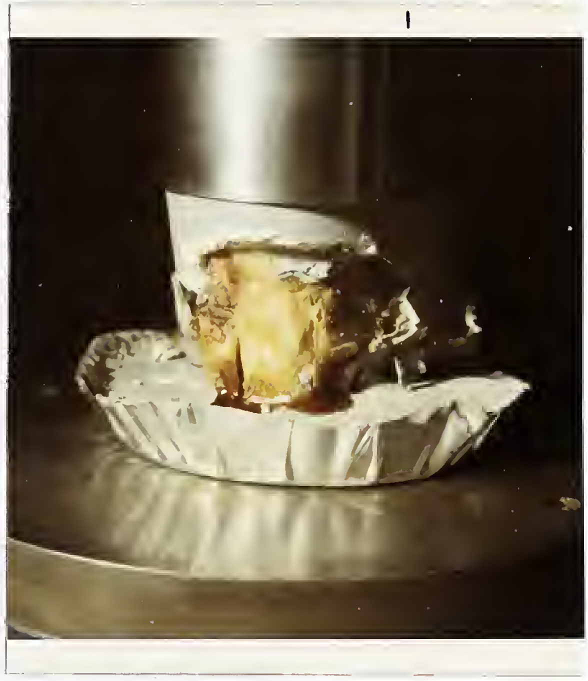
C. At 300 kiloponds

D. At failure point, 340 kiloponds

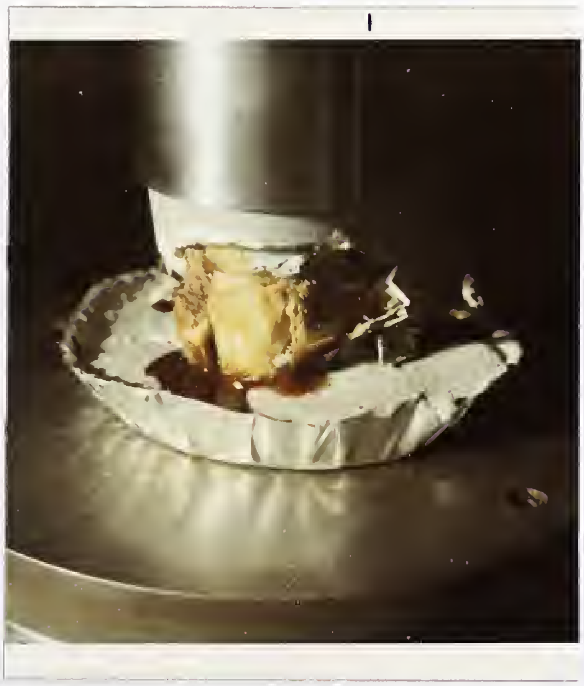
Fig. 12. Type II Graft-Vertebrae Construct Under Compression



A. At 100 kiloponds



B. At 200 kiloponds



C. At failure point,
250 kiloponds

Fig. 13. Type III Graft-Vertebrae Construct Under Compression

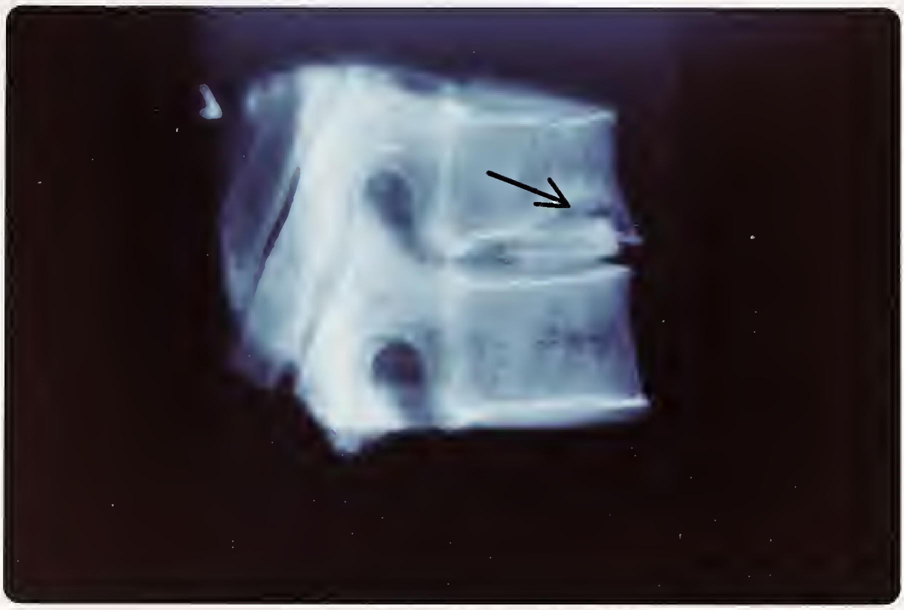


Fig. 14. Type I Graft-Vertebrae Construct Post-Compression Arrows Demonstrate Graft Violating Vertebral End-Plate.

Fig. 15. Longitudinal Section Type I. Arrow Points to Failure of Upper Vertebral End-Plate.

Fig. 16. Longitudinal Section Type I. No Evidence of Failure Post-Compression.

Anterior View

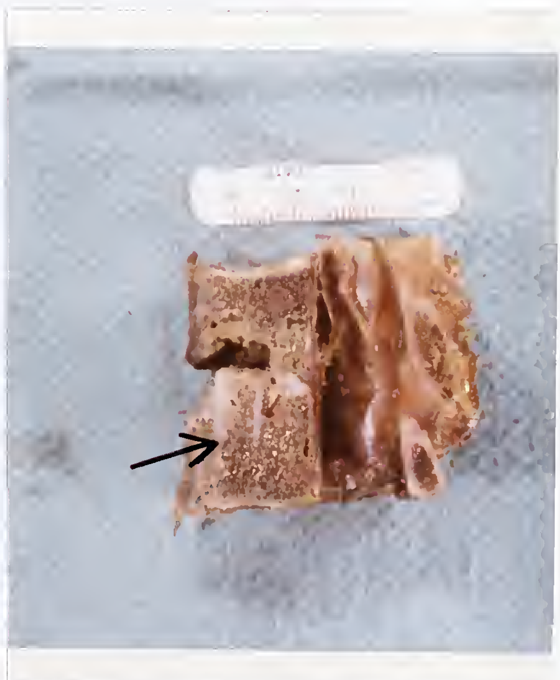


Fig. 17. Type II Construct. Arrows Show Graft Driven into Cancellous Bone of Lower Vertebra.

Fig. 18. Type II Construct Longitudinal Section Reveals Destruction of Cancellous Bone of Upper Vertebral Body

Fig. 19A. Type III Construct. Arrow shows widening of lateral margin of graft site. Graft has also been driven into cancellous bone of lower vertebra.

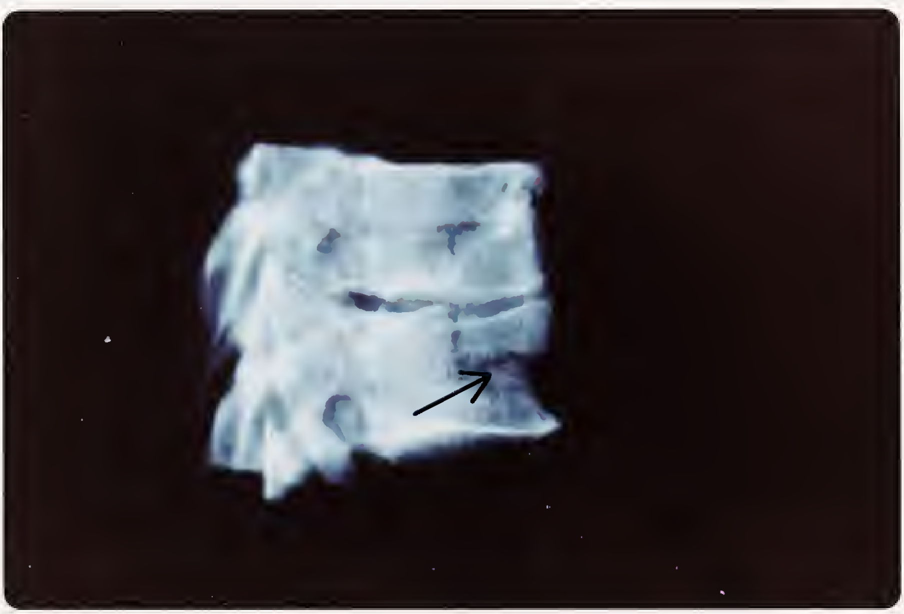


Fig. 19B. Type III Construct. Arrow shows destruction of cancellous bone below graft.

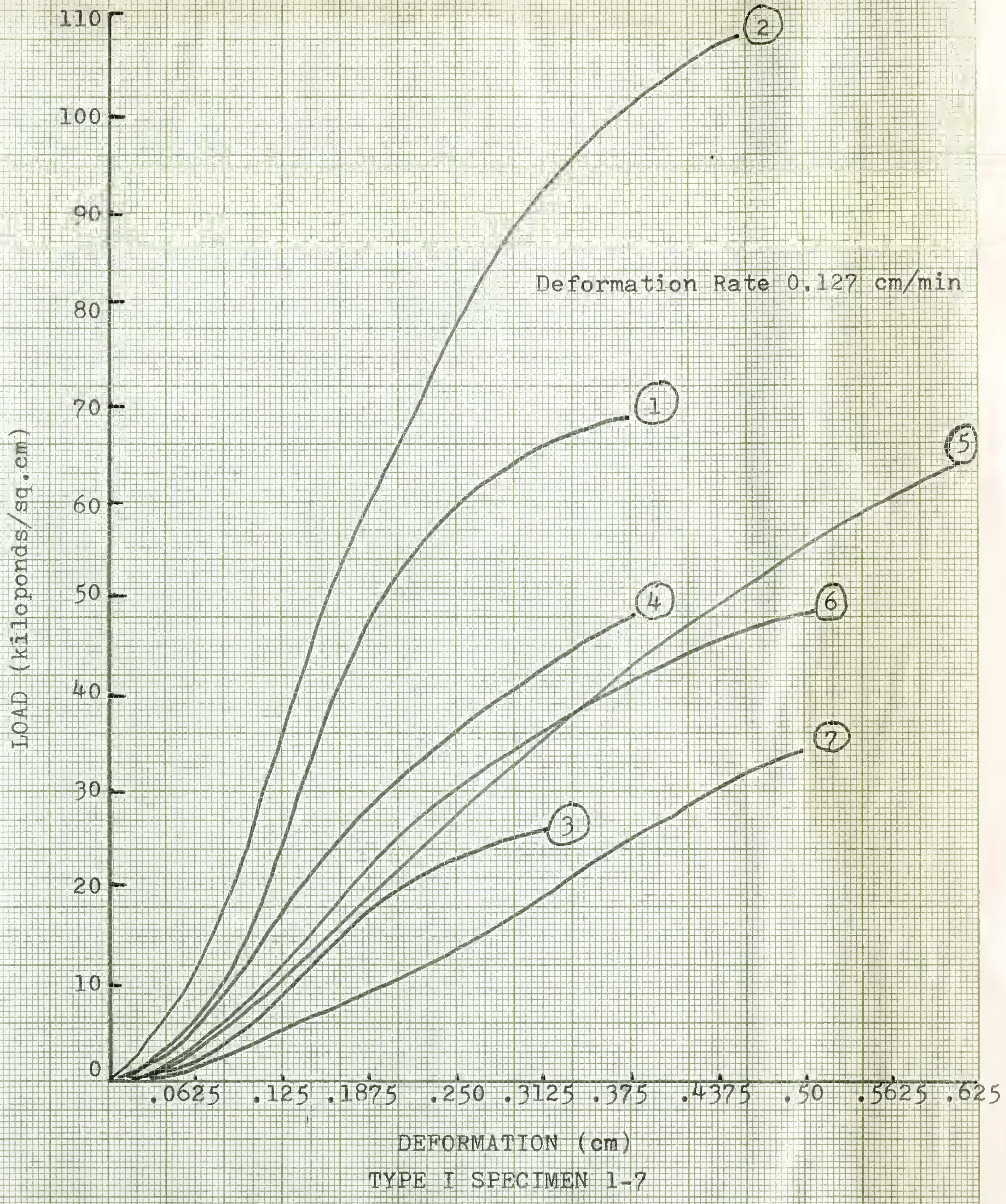


FIG. 20. LOAD-DEFORMATION CURVE
TYPE I GRAFT-VERTEBRAE CONSTRUCT

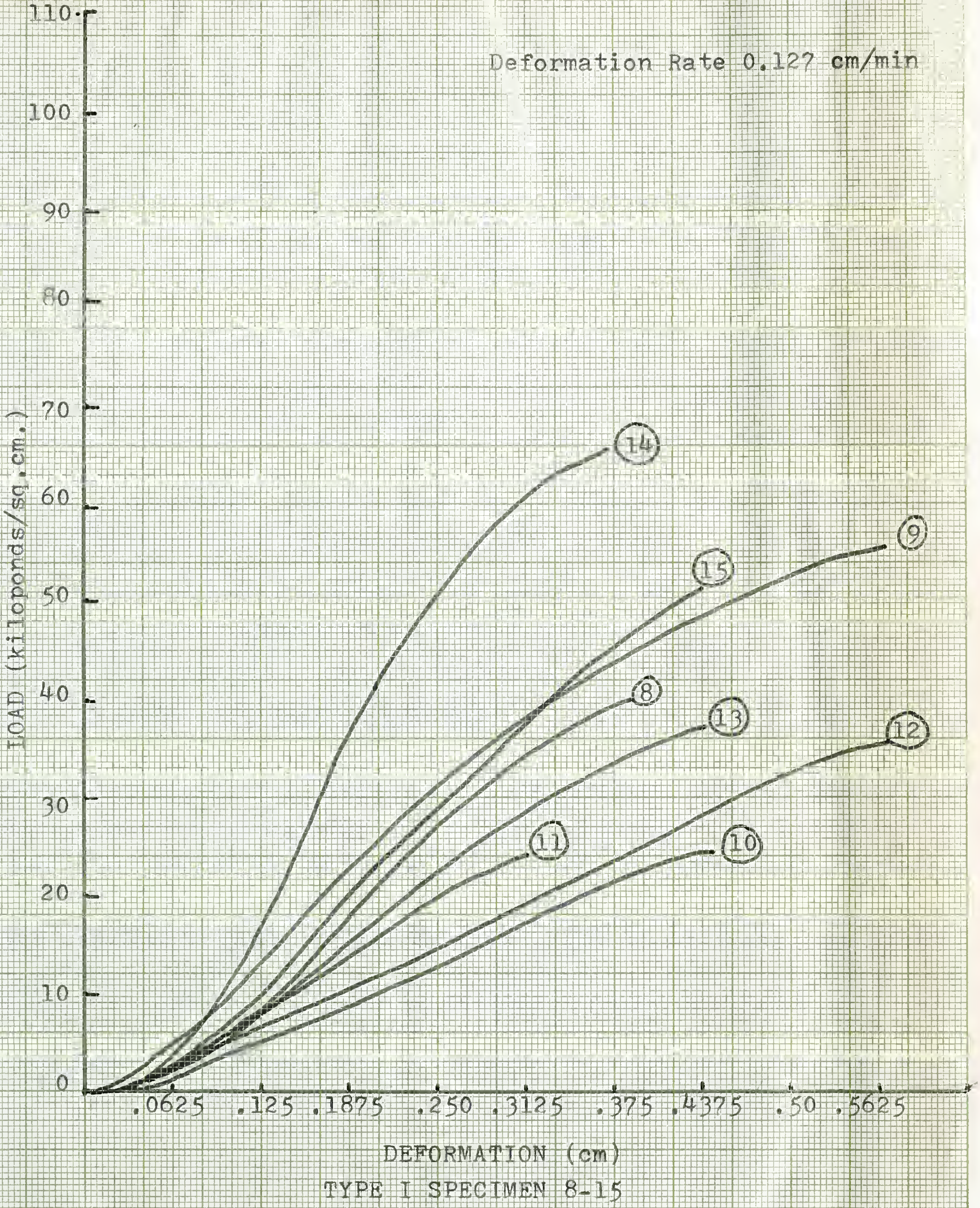


FIG. 21. LOAD-DEFORMATION CURVE TYPE I GRAFT-VERTEBRAE CONSTRUCT

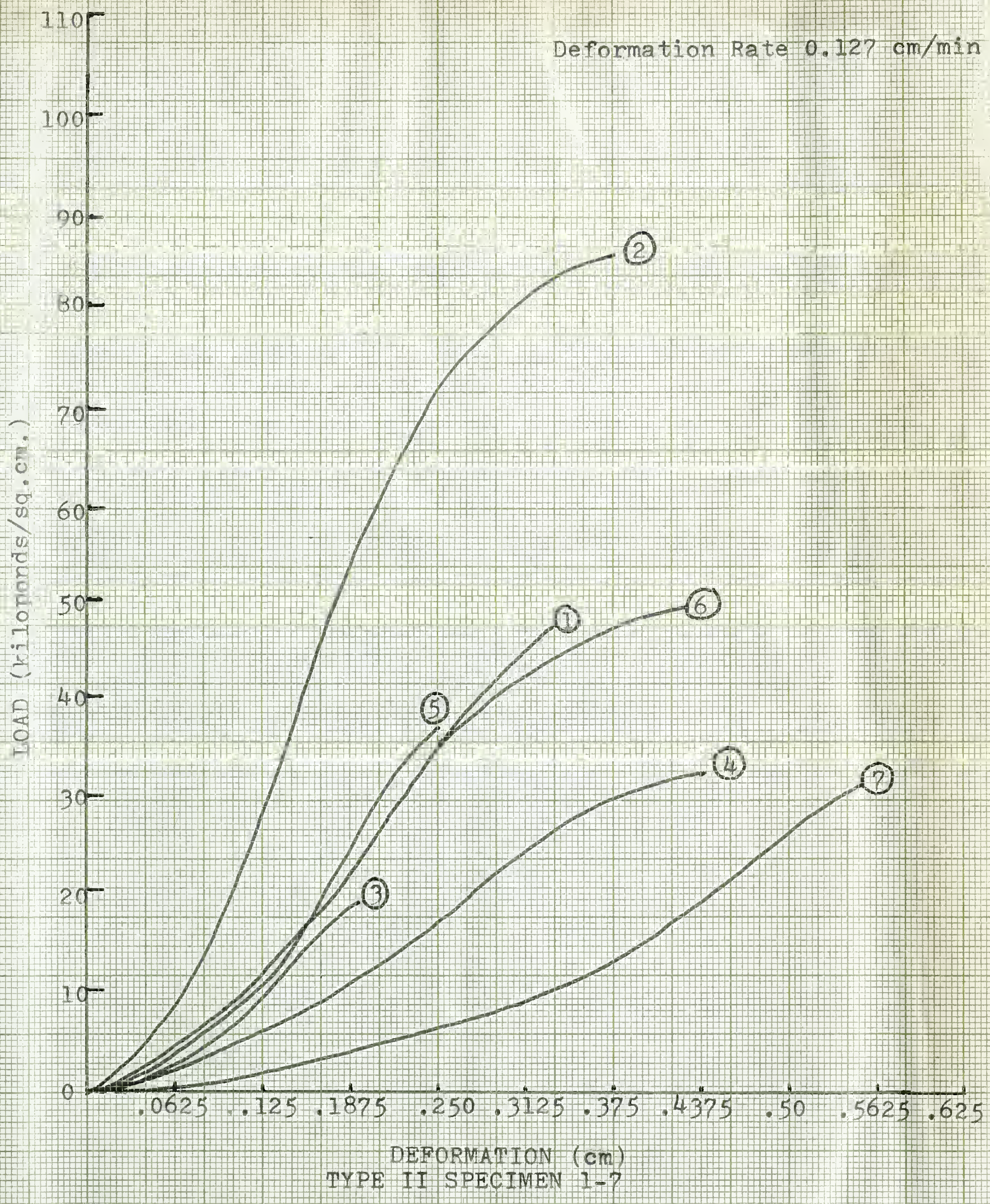
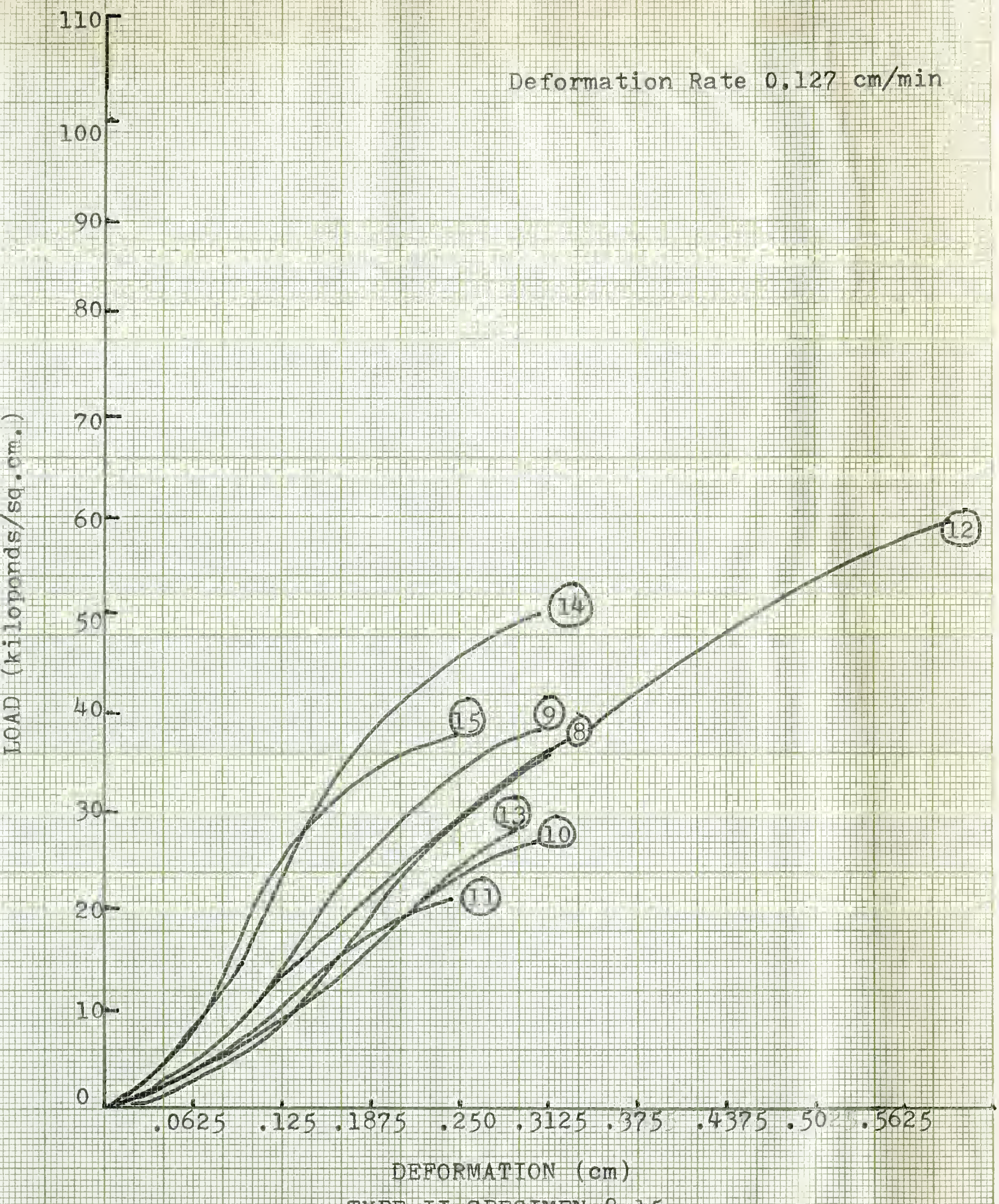


FIG. 22. LOAD-DEFORMATION CURVE
TYPE II GRAFT-VERTEBRAE CONSTRUCT



TYPE II SPECIMEN 8-15

FIG. 23. LOAD-DEFORMATION CURVE TYPE II GRAFT-VERTEBRA CONSTRUCT

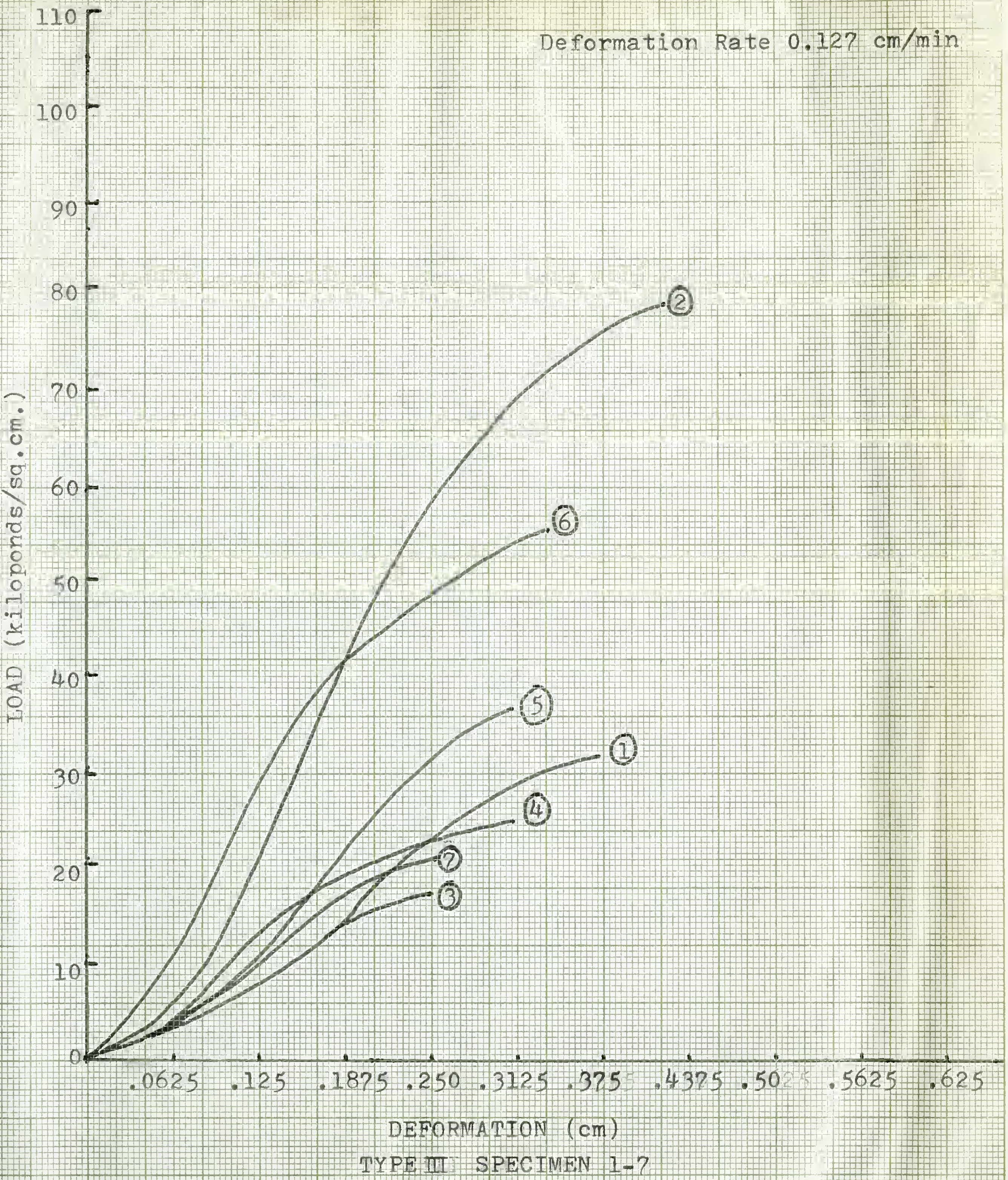


FIG. 24. LOAD-DEFORMATION CURVE
TYPE II GRAFT-VERTEBRAE CONSTRUCT

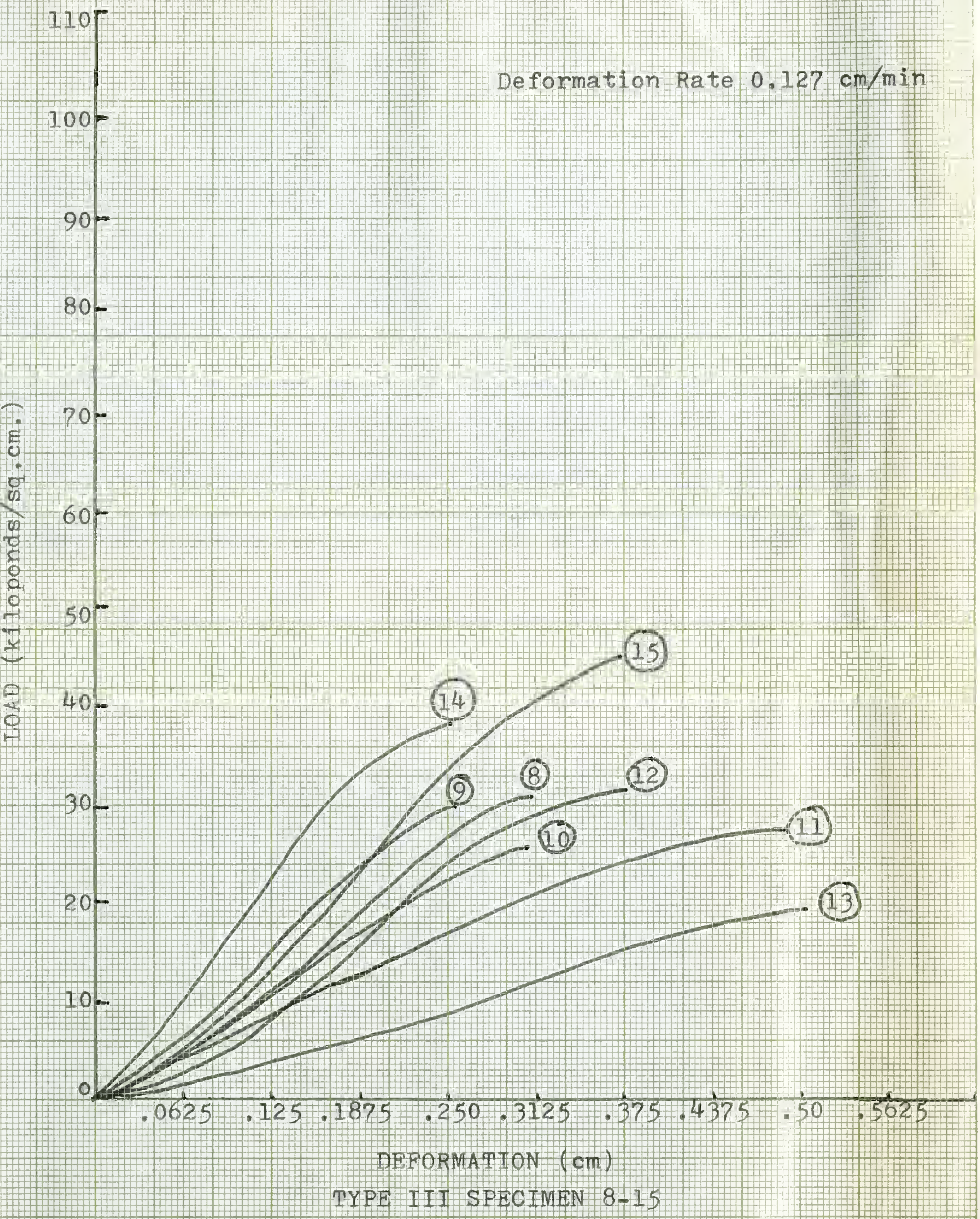


FIG. 25. LOAD DEFORMATION CURVE
TYPE III GRAFT-VERTEBRAE CONSTRUCT

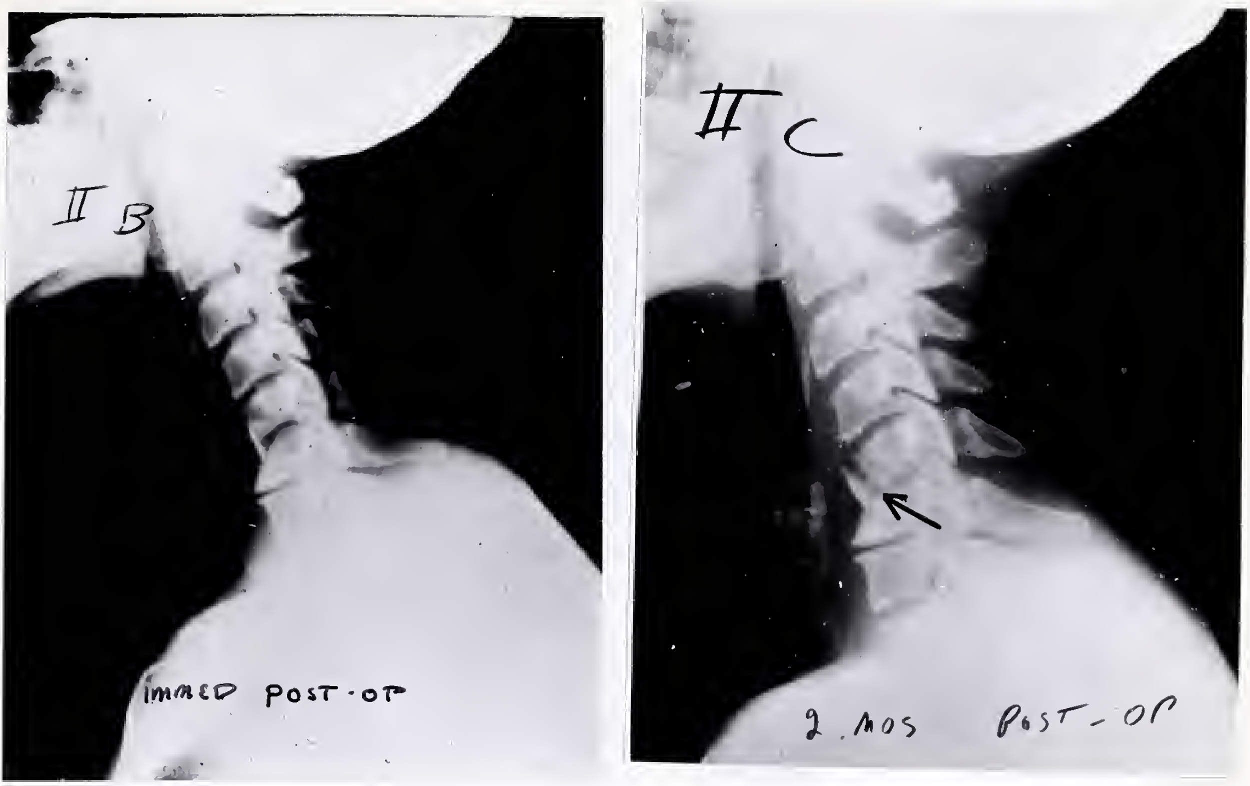


Fig. 26. Roentgenograms from Kebish and Keggi (1967) demonstrating anterior collapse of the vertebral body on the cortical dowel graft.

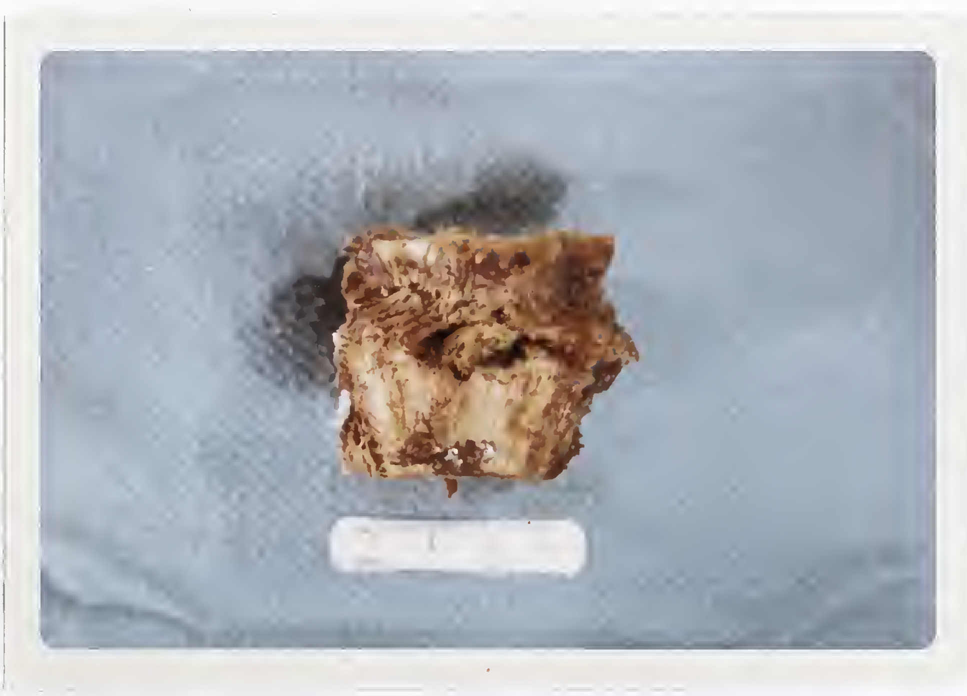


Fig. 27. Anterior view of Type II Construct, showing anterior collapse of the vertebral bodies over the graft.

Fig. 28. Roentgenogram from Kebish and Keggi (1967), demonstrating non-parallel cortical margins of the dowel graft.

Fig. 29. Roentgenogram of Type II Construct, also demonstrating non-parallel cortical margins of dowel graft.